ARL-MR-0934 • JULY 2016

US Army Research Laboratory

A Computational Approach for Automated Posturing of a Human Finite Element Model

by Justin McKee and Adam Sokolow

NOTICES

Disclaimers

ARL-MR-0934 ● July 2016

US Army Research Laboratory

A Computational Approach for Automated Posturing of a Human Finite Element Model

by Justin McKee
Bennett Aerospace, Inc., Cary, NC

Adam Sokolow
Weapons and Materials Research Directorate, ARL

REPORT DOCUMENTATION PAGE	*Form Approved* *OMB No. 0704-0188*

Public reporting burden for this collection of information is estimated to average 1 hour per response, including the time for reviewing instructions, searching existing data sources, gathering and maintaining the data needed, and completing and reviewing the collection information. Send comments regarding this burden estimate or any other aspect of this collection of information, including suggestions for reducing the burden, to Department of Defense, Washington Headquarters Services, Directorate for Information Operations and Reports (0704-0188), 1215 Jefferson Davis Highway, Suite 1204, Arlington, VA 22202-4302. Respondents should be aware that notwithstanding any other provision of law, no person shall be subject to any penalty for failing to comply with a collection of information if it does not display a currently valid OMB control number.

PLEASE DO NOT RETURN YOUR FORM TO THE ABOVE ADDRESS.

1. REPORT DATE *(DD-MM-YYYY)*	**2. REPORT TYPE**	**3. DATES COVERED (From - To)**
July 2016	Memorandum Report	October 2015–October 2016

4. TITLE AND SUBTITLE A Computational Approach for Automated Posturing of a Human Finite Element Model	**5a. CONTRACT NUMBER**
	5b. GRANT NUMBER
	5c. PROGRAM ELEMENT NUMBER
6. AUTHOR(S) Justin McKee and Adam Sokolow	**5d. PROJECT NUMBER** FPSP31
	5e. TASK NUMBER
	5f. WORK UNIT NUMBER
7. PERFORMING ORGANIZATION NAME(S) AND ADDRESS(ES) US Army Research Laboratory ATTN: RDRL-WMP-B Aberdeen Proving Ground, MD 21005-5066	**8. PERFORMING ORGANIZATION REPORT NUMBER** ARL-MR-0934
9. SPONSORING/MONITORING AGENCY NAME(S) AND ADDRESS(ES)	**10. SPONSOR/MONITOR'S ACRONYM(S)**
	11. SPONSOR/MONITOR'S REPORT NUMBER(S)

12. DISTRIBUTION/AVAILABILITY STATEMENT

Approved for public release; distribution is unlimited.

13. SUPPLEMENTARY NOTES

contact author's email: <adam.c.sokolow.civ@mail.mil>.

14. ABSTRACT

Human body models for use in Soldier protection are fundamental building blocks. A major gap in our current human body modeling capability is the proper accounting of posture. Posture relates to Soldier protection by influencing the path that loading will be transferred into the body and is a major source of variability. The development of a finite element (FE) model is a time consuming and labor intensive process and taking the approach of developing multiple models to address the range of postures relevant to each threat is not realistically feasible. Here we discuss the development of a tool to automate the process of positioning an existing FE model of the leg, lumbar spine, and thoracic spine into a desired posture. This approach allows the repeated use of a single FE model to address multiple problems and provides an automated and consistent method that produces a range of FE meshes to study how initial posture might relate to injury.

15. SUBJECT TERMS

posture, human body, finite element, leg, spine

16. SECURITY CLASSIFICATION OF:			**17. LIMITATION OF ABSTRACT**	**18. NUMBER OF PAGES**	**19a. NAME OF RESPONSIBLE PERSON** Adam Sokolow
a. REPORT Unclassified	**b. ABSTRACT** Unclassified	**c. THIS PAGE** Unclassified	UU	42	**19b. TELEPHONE NUMBER (Include area code)** 410-306-2985

Standard Form 298 (Rev. 8/98)
Prescribed by ANSI Std. Z39.18

Contents

List of Figures

List of Tables

INTENTIONALLY LEFT BLANK.

1. Introduction

Predicting occupant injuries that result from an underbody blast event with finite element (FE) models of the human body present many numerical challenges. During such an event, explosive gases and soil impart momentum to the vehicle hull and subsequently impart accelerations to the floor plate and vehicle occupants. Accelerative injuries, a term typically used in automotive literature, is commonly used to describe these injuries. However, this nomenclature is misleading in that it combines shock-induced injuries with those that result from structural perspective of accumulated strain. The model detail, resolution, and numerical techniques used to solve for the mechanics in each case is quite different. The current strategy is to adequately predict injuries due to accumulated strains on the load-bearing structures of the body. This type of model can be generated for the entire body and can have a more immediate impact on improvements in Soldier protection. However, the methods developed here are general and can be applied rapidly to models that address other physics.

A commonly used posture when modeling the mounted Solder is the 90-90-90 posture where each number indicates the measured angle in degrees between the hip and femur, femur and tibia, and tibia and foot, respectively. While this posture represents a common and simple seated position, it is unlikely that it will reproduce the wealth of injuries seen in theater where the injuries can be combinations of structural kinematic failures and shock induced. Different seated postures and the angle of the leg relative to the floor of the vehicle will alter the load path of the leg. Investigating the effect of posture during an under body blast event will help to illustrate ranges of seating positions that may increase or decrease risk of injury, the influence of personal protective equipment (PPE) on extremity kinematics, and accumulated strain-type injuries.

Accounting for posture in a FE simulation is a challenge when modeling the human body. Not only does posture act as a source of variability, it also requires modification of a FE mesh, which is already a non-trivial process for the human and other complex structures and represents a large time investment even for minor changes. The common methods to account for posture that are employed by the research field include the following: obtaining source geometries in the posture being tested, a so-called posturing “by hand” where geometries are moved to what “looks correct” and

sculpted and smoothed to account for any abnormalities, and picking points on the structure and simulating the motion through load curves.

Each of these methods has its merits. Posturing by hand produces a quick source geometry from a pre-existing one; however, the accuracy and biofidelity of the joint motion is then subjective and subsequent meshing of the new geometry is still required. Obtaining medical scans in various postures solves the biofidelity/accuracy issue but requires specialized hardware and brings with it extensive medical image processing to convert images into surface and volume definitions of the structures. Following this is a time consuming meshing effort, that must be performed for each new scan (this too is an active research subject). Both of these approaches, however, make the total number of geometries that need to be meshed a combinatorial problem. This makes it no longer feasible to use a brute force approach and computer-automated meshing procedures along with automated computer vision are necessary.

The load curve approach avoids repeated mesh generation and passes the time consuming step to a computer simulation. This approach is often taken to make minor alterations by hand where one part is rotated and translated in a simple manner. Posturing the human using load curves, however, presents a biofidelic challenge since the joints in the body, although commonly thought of in simple terms like "hinges" and "ball and socket", rarely move as perfect hinges or ball and socket joints. This can become an issue when 2 stiff structures like bone are connected by soft structures and an over-determined motion from assuming a hinge joint results in pinching, binding, and or rupturing of neighboring soft tissues. Other issues include properly accounting for the change in length of a muscle during flexion or extension that result from different postures.

The goal of the work presented here is to develop an automated method to reposition any existing FE model of the leg and spine into different postures using biofidelic descriptions of the joint motions. This type of architecture reduces the burden of mesh development. Mathematical descriptions of the relative motion of bones at the joints are used to define a motion that is carried out through a FE simulation. The motion of the bones pulls the attached soft tissues into place to form the new posture. While the functions developed here are specific to the leg and spine, the overall architecture can readily be generalized to the remainder of the body. This

report is organized as follows: first we present a general overview of the posturing procedure, then we discuss specific joint functions and provide more detail for each joint considered, and finally we discuss the assembly process, its conversion to a FE simulation load curve, and some preliminary results of these simulations for the lower leg.

2. Overview of the Posturing Procedure

The posturing procedure is divided into 4 steps. The first step is a sub-division of the FE model into relevant relationship sets. The second step involves an application of a linear transformation to objects within a relationship set. This linear transformation, or *joint function*, maps a set of coordinate axes from a reference configuration to the postured configuration. The third step re-assembles the FE model by combining each of the local transformations applied to the sets. The fourth step is a load curve generation step that uses time-parameterized variables and repeated applications of the second and third steps. These procedures are described for the right side of the body in this report, but the posture process has an option to be applied to the left side of the body as well.

An overview of steps 2 and 3 are presented in Fig. 1 using the foot and tibia as an example. Figure 1a shows the information stored by each bone in the procedure. First, local coordinate systems are defined for each of the bones that are used in the posture procedure. Each bone stores its local coordinate system origin and axis directions under the global coordinate system. The local coordinate system is then used to store the node coordinates for the mesh and the origin and axis directions of neighboring bones' coordinate systems. Next, a joint function prescribes the rigid-body motion of the 2 bones. Figure 1b shows an example of storing the neighboring coordinate systems based on the relative motion. These functions are shown in detail in later sections. The next step shown in Fig. 1c assimilates the relative motions of the joints into a description in a global coordinate system. This is prescribed for a set of joints where the cumulative effect constitutes whole body posturing. Thus, the rigid-body motion of the entire skeleton can be prescribed. This procedure is done for all of the nodes in the bones and results in a set of load curves that can be read into a FE simulation. The simulation moves the bones as rigid bodies and allows the softer tissues to conform to the motion according to their constitutive response. The final location of the nodes represents the postured model.

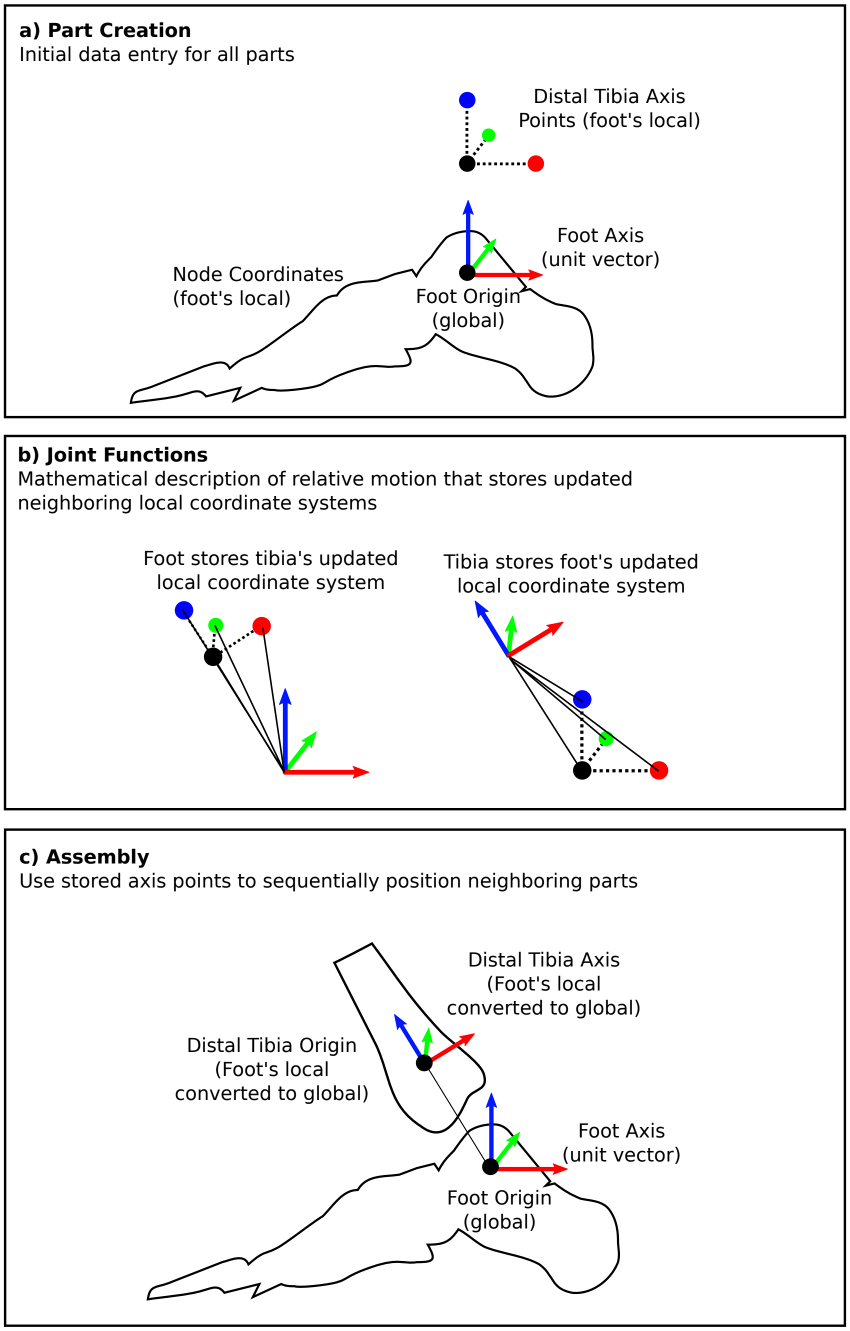

Fig. 1 An overview of the posture procedure. a) Bones store information about their local coordinate system, node coordinates, and neighboring local coordinate systems. b) Joint functions use a mathematical description of relative motion to store updated local coordinate systems. c) The stored local coordinate systems are used to assemble the bones of the leg in the global coordinate system.

The approach is flexible in that improvements to the joint functions can be made at any time. Thus, at first a model might assume a perfect hinge and then later be enhanced/corrected to allow for a sliding hinge, or other more complicated joint functions. This enhancement can improve the result without forcing any re-write of the architecture, or any re-meshing of geometries. Since the final posture depends in part on the underlying FE model, the approach also benefits from any improvements made to the model. The current procedure has been successfully used to adjust the ankle, knee, and hip angles of the leg model described in previous work[1] and the lumbar and thoracic vertebrae. The posture procedure allows the leg to transition from a standing to a seated position, thus expanding the range of threats that the initial model can be used to simulate.

3. Joint Functions

A joint function can be thought of as a combination of linear transformations applied to neighboring bones that prescribes their relative motion. During the posturing procedures, the bones are treated as rigid bodies, thus only the transformation of a local coordinate system is needed as a reference for determining the nodal locations in the global system.

In the following text, vectors are denoted using bold letters, with the exception of unit vectors, which may appear in regular weight font but with a hat to indicate they have been normalized. The superscript on a vector indicates the coordinate system its components are given in terms of, and the subscript indicates the object the vector points to (in some cases the subscript is omitted due to redundancy of the notation). The mathematical description of the joint functions and the assembly process is simple vector operations, but due to operations taking place in local coordinate systems the language describing these procedures can be quite complicated. The example and figure given are in 2 dimensions, but the extension to 3 dimensions is trivial. Figure 2a shows 2 local coordinate systems, one for the femur (superscript f) and one for the tibia (superscript t). Although we think of the 2 coordinate systems separately, their components are equal since they are given in their own reference frames (e.g, $\hat{x}^f = \hat{x}^t$ and $\boldsymbol{O}^t = \boldsymbol{O}^f$), although if we were to express the location of the origin of the tibia system $\boldsymbol{O}^t$ in the femur coordinate system to get $\boldsymbol{O}_t^f$ the components of these 2 vectors would not be equal. The third set of axes (dashed lines) represent the application of a joint function, symbolically

indicated by the green arrow, which rotates and translates the system. Vector $\boldsymbol{n}^t$ represents the location of a node of the tibia mesh in the tibia coordinate system, and $\boldsymbol{n}^{t'}$ represents the same node in the rotated and translated system. The components of $\boldsymbol{n}^t = \boldsymbol{n}^{t'}$ since both are expressed in their own local coordinate systems. This observation is part of the mesh treatment that simplifies the number of rotations that must be applied to a single node in the mesh even if it is subjected to multiple joint functions.

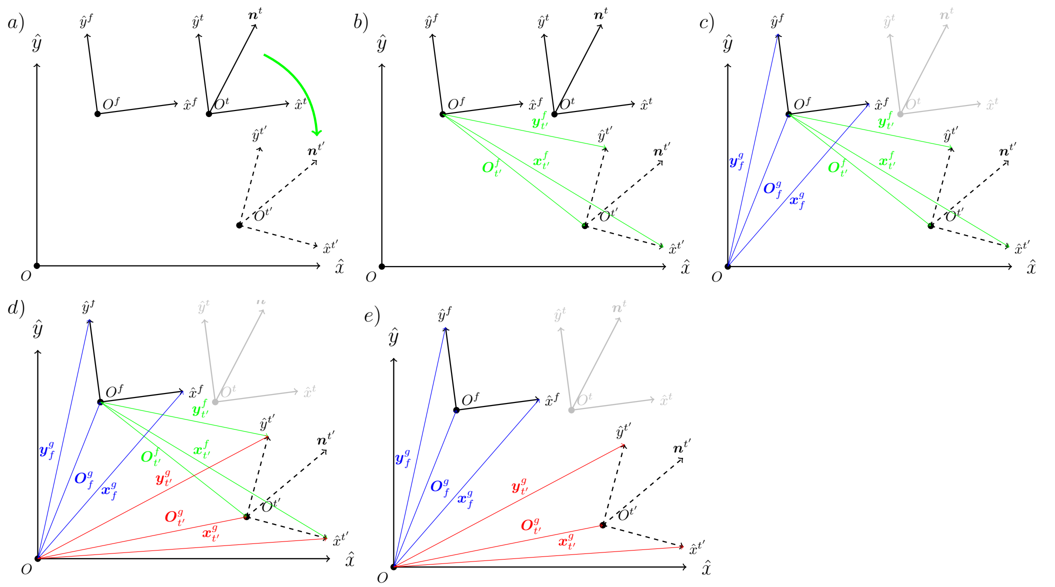

Fig. 2 a) Joint function (green) is a linear transformation of the local tibia coordinate system to its primed location. b) Establishment of reference vectors from the femur coordinate system to the primed tibia system. c) Establishment of a global reference frame. d) The relevant vectors of the primed tibial location in the global frame. e) The complete description of a vector n in the new system.

Figure 2b introduces 3 vectors that are used to store the information that specifies the result of applying the joint function to the tibia coordinate system. These are indicated by the green arrows and the vectors are given in terms of the femur coordinate system and give the location of the primed tibia origin $\boldsymbol{O}^f_{t'}$, the primed x-hat vector $\boldsymbol{x}^f_{t'}$, and primed y-hat vector $\boldsymbol{y}^f_{t'}$. These 3 vectors are all that is needed to reconstruct the primed-tibia system relative to the femur system.

Figure 2c introduces some notion of a reference frame, in this example it is the anchor point from which to assemble the femur-tibia system, but in general could be part of another bone in the assembly. In the case of the example, this reference frame already existed as soon as we drew our tibia and femur coordinate systems

separate from one another, so in essence we have the freedom to choose how the femur is placed in this system. This choice establishes the femur vectors expressed in the global coordinate system (blue arrows), giving the location of the femur origin $\boldsymbol{O}_f^g$, the location of the x-hat vector for the femur system $\boldsymbol{x}_f^g$, and the y-hat vector $\boldsymbol{y}_f^g$.

Once a reference system has been established, simple tail-to-head addition of the joint function vectors (green arrows) with the femur origin (blue arrows) yields the primed-tibia coordinate system relative to the global system (red arrows) (Fig. 2d). In other words,

$$\boldsymbol{O}_{t'}^g = \boldsymbol{O}_f^g + \boldsymbol{O}_{t'}^f \,, \quad \boldsymbol{x}_{t'}^g = \boldsymbol{x}_{t'}^f - \boldsymbol{O}_f^g \,, \text{ and } \quad \boldsymbol{y}_{t'}^g = \boldsymbol{y}_{t'}^f - \boldsymbol{O}_f^g \,. \tag{1}$$

From these vectors one can calculate the direction vectors of the primed-tibia system in global coordinates and convert a node given in terms of the tibia coordinate system to the global system. This is done through a rotation matrix given by

$$\boldsymbol{R}_{t'} = \left[(\boldsymbol{x}_{t'}^g - \boldsymbol{O}_{t'}^g)^T \quad (\boldsymbol{y}_{t'}^g - \boldsymbol{O}_{t'}^g)^T \right] \,, \tag{2}$$

such that a node in the tibia coordinate system $\boldsymbol{n}^t$ expressed in the postured-global system is simply

$$\boldsymbol{n}_{t'}^g = \boldsymbol{R}_{t'}^T \boldsymbol{n}^t + \boldsymbol{O}_{t'}^g \,, \tag{3}$$

which is shown in Fig. 2e and a similar operation on the femur yields

$$\boldsymbol{R}_f = \left[\left(\boldsymbol{x}_f^g - \boldsymbol{O}_f^g\right)^T \quad \left(\boldsymbol{y}_f^g - \boldsymbol{O}_f^g\right)^T \right] , \tag{4}$$

and a node in the local femur coordinate system placed in the global system is

$$\boldsymbol{n}_f^g = \boldsymbol{R}_f^T \boldsymbol{n}^f + \boldsymbol{O}_f^g \,. \tag{5}$$

3.1 Ankle

The ankle is currently defined as an idealized ball and socket joint. A local coordinate system is defined for the foot whose origin is at the center of the talus. A local coordinate system for the distal tibia is defined directly superior to the foot origin at the base of the tibia. The axes of the distal tibia are parallel with the axes of the foot in the reference posture. The rigid-body motion of the tibia is taken as a pure rotation about this origin. This pure rotation constitutes the ankle function. The current function supports inputs of dorsiflextion-plantar flexion and inversion-eversion.

Figure 3 shows an illustration of the ankle function. Dorsiflexion-plantar flexion is measured as the angle that the inferior-superior axis of the tibia makes when projected onto the sagittal plane (θ). Inversion-eversion is measured as the angle that the inferior-superior axis of the tibia makes when projected onto the coronal plane (ϕ). The foot is fixed and the tibia is rotated to achieve the target posture.

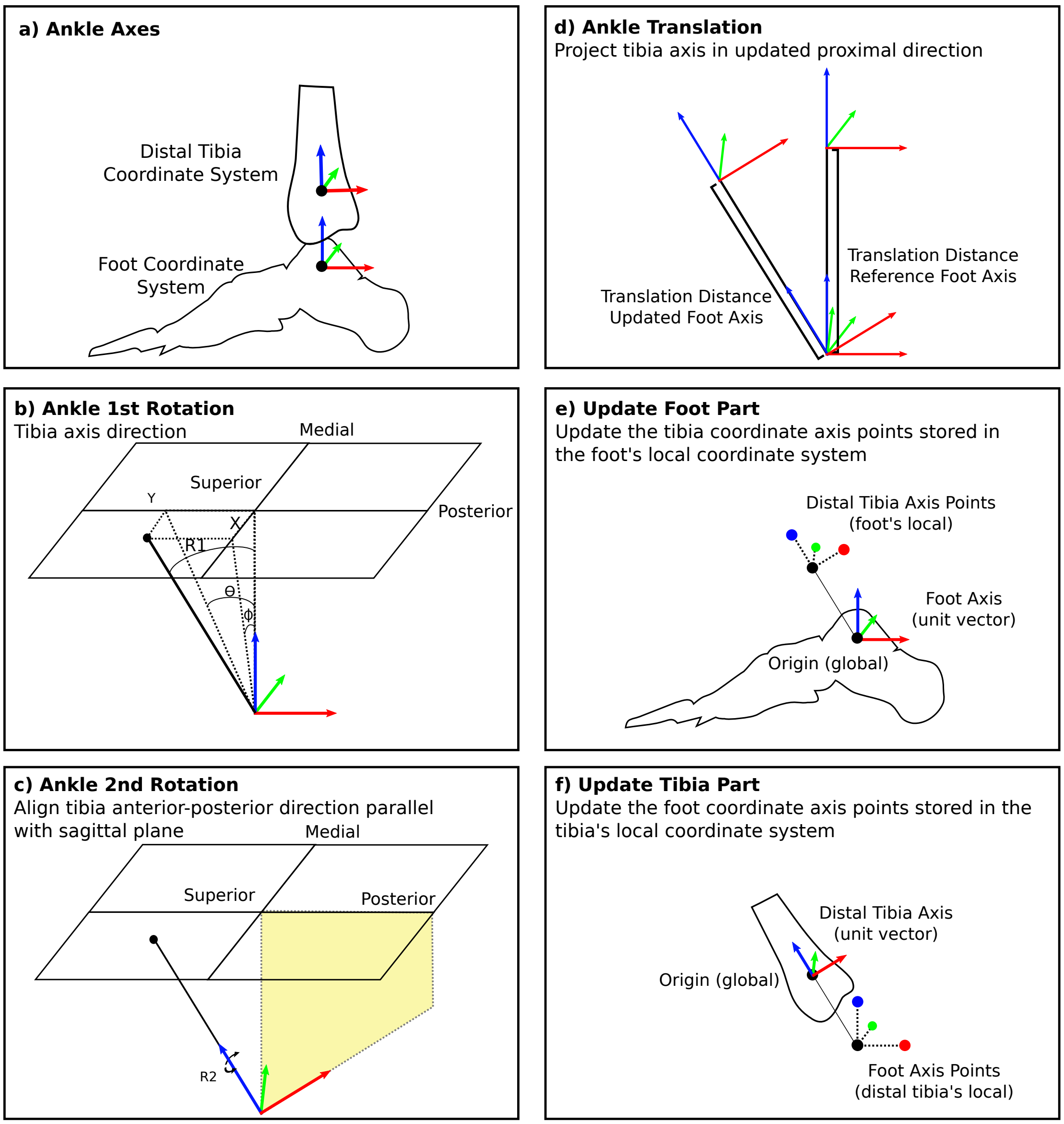

Fig. 3 The ankle is defined as a ball and socket joint. The tibia coordinate system rotates about the foot based on input anatomical angles.

Two rotations are performed to reach the target posture. The first rotation is shown in Fig. 3b. A point and vector that form the desired angles are calculated first. A rotation is found to align the inferior-superior axis of the tibia with this new vector. This rotation is applied to the tibia's coordinate axis to complete the first rotation. The second rotation is shown in Fig. 3c. The tibia's anterior-posterior axis is aligned parallel with the sagittal plane. This second rotation is performed about the inferior-superior axis of the tibia after applying the previous rotation.

The updated location of the tibia's origin is determined using the updated axis directions. This procedure is illustrated in Fig. 3d. The origin is projected from the foot's origin along the updated inferior-superior axis by the same distance that separated the foot and tibia axis in the reference posture.

The last step is to store the updated coordinate systems under their neighbor's coordinate system. This is shown in Fig. 3e for the foot and Fig. 3f for the tibia. This step stores the rotation by saving the relative position of the foot and tibia and is equivalent to the generalized description in Fig. 2b. Either part can be plotted in the posture determined from this function if the coordinate system of the other is known.

3.2 Knee

The relative motion of the tibia, femur, and patella are defined based on kinematic data described by Li et al.[2] Li et al. uses a dual-orthogonal fluoroscopy system to measure the translation, rotation, and tilt of the femur and patella relative to the flexion angle. The posture procedure makes use of this coupling to define the full kinematics of the knee using a simplified single input of a flexion angle.

Positioning of the knee is illustrated in Fig. 4. Figure 4a shows the axes that are used in this process. The transepicondylar axis (TEA) is defined as the line that connects the medial and lateral collateral ligament attachment sites. The femur's distal coordinate system origin is placed at the mid point of this line. The long axis of the femur originates from the distal femur coordinate origin along the length of the femur's shaft. The TEA and long axis of the femur are not aligned with the axes of the femur's local coordinate system and are not perpendicular with each other. These 2 axes are used to find a rotation matrix that is then applied to the femur's coordinate system. Their relative orientation will remain the same throughout the

procedure. The tibia's proximal coordinate system has the origin contained within the interior of the tibia directly inferior to the femur's coordinate system. The tibia and its local coordinate system are held fixed during the knee-positioning process. Then the patella's local coordinate system is defined at the center of the bounding box that contains the patella geometry.

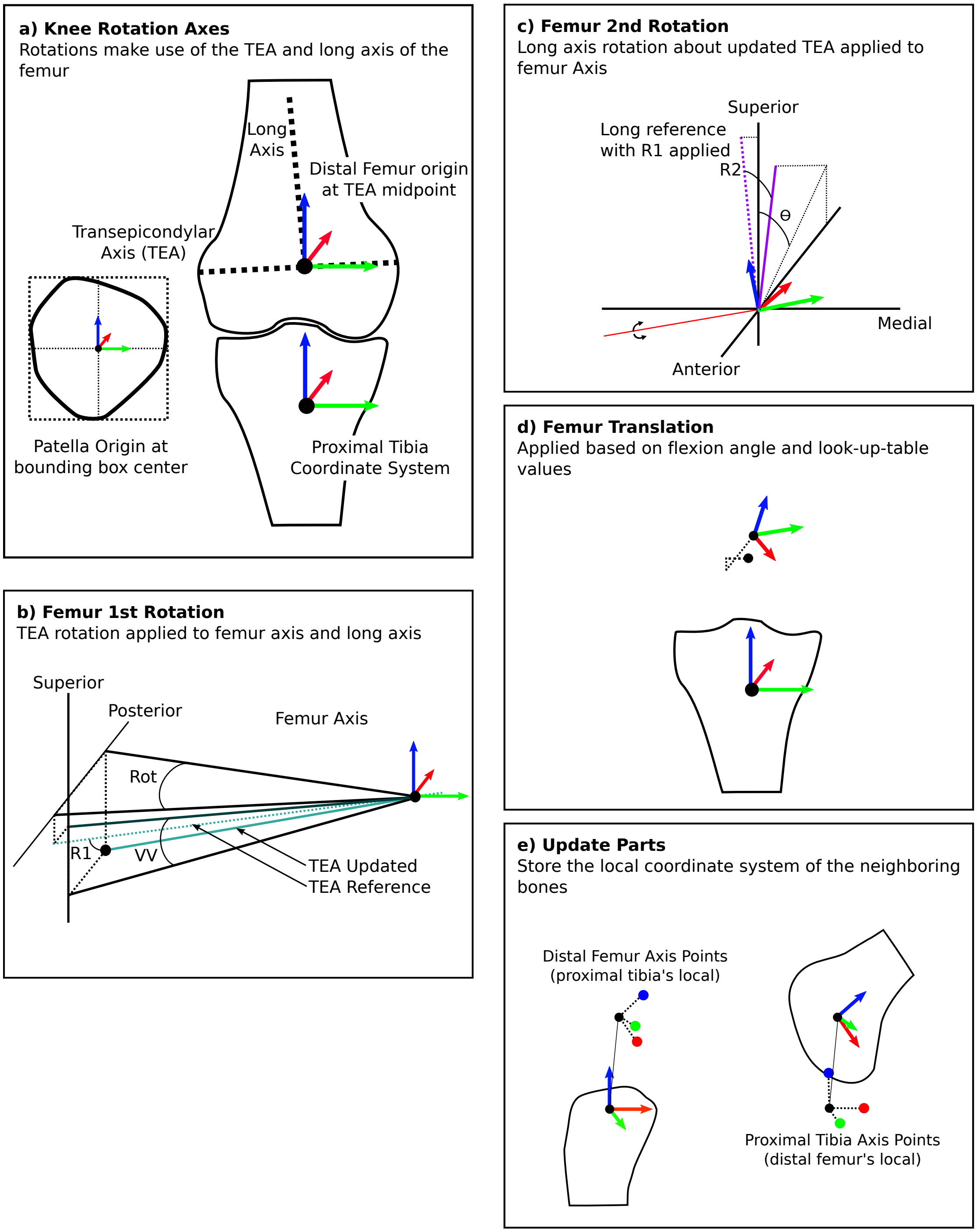

Fig. 4 The motion of the femur and patella are defined by kinematic data related to the degree of knee flexion

Rotation is applied to the femur and patella based on the flexion angle. Additional parameters that correspond with the current flexion angle are determined from the interpolation of a lookup table using values derived from the data reported by Li et al.[2] with some deviation away from their average values to account for the geometry of the knee used in the model. Rotation of the knee and patella takes place in 2 steps. The first rotation is illustrated for the femur in Fig. 4b. This first step accounts for internal-external rotation and varus-valgus rotation. Internal-external rotation of the femur is measured as the angle between the current TEA and the reference TEA when projected to the transverse plane (Rot). Varus-valgus rotation is measured as the angle between the current TEA and the reference TEA when projected to the coronal plane (VV). A point and vector are found that will provide the target angles, and a rotation is calculated to orient the TEA in this direction. This rotation is then also applied to the femur local coordinate system and the long axis.

The second rotation applies the flexion angle and is shown in Fig. 4c. Flexion is defined as the angle between the current long axis and the reference long axis when projected to the sagittal plane (theta). The long axis and the femur's coordinate system are rotated about the updated TEA until the target flexion angle is reached.

Femur translation illustrated in Fig. 4d is taken into account from a lookup table (see Appendix). An illustration of the translation is shown in Fig. 4e. The translation is applied to the origin of the femur's local coordinate system using the axis directions of the reference tibia coordinate system.

A similar positioning process is applied to the patella. The first rotation makes use of the patella's medial-lateral axis in place of the TEA to apply internal-external rotation and tilt. Internal-external rotation is the angle of rotation projected into the transverse plane, and tilt is the angle of rotation projected into the coronal plane. These angles are measured as the angle between the reference medial-lateral axis to the updated medial-lateral axis. Flexion is then applied to the patella by rotating the coordinate system about the updated medial-lateral axis. Flexion is defined as the angle between the reference and updated inferior-superior axis projected onto the sagittal plane. Translation is applied to the patella using the lookup table values and the same method as the femur translation.

The last step is to store neighboring coordinate systems to preserve the rotation when creating the assembly. The tibia part stores both the distal femur and patella

coordinate system locations in the tibial local coordinate system. The femur and patella store the location of the tibia's coordinate system location within their own local coordinate systems. The patella can became a passive component with motion determined by its soft tissue attachments in the later FE step by simply not running the patella portion of the knee function. This capability helps to demonstrates the flexibility of the posture procedure's generalized architecture.

3.3 Hip

The hip posture function is shown in Fig. 5. The hip follows a similar procedure to that of the ankle and assumes the joint is adequately described by a ball and socket joint. Figure 5a shows the initial location and orientation of the femur and pelvis local coordinate systems. The proximal femur local coordinate system is located in the approximate center of the femoral head and serves as the center of rotation for the femur. The origin of the pelvis local coordinate system is in the same location as the femur. There is no relative translation between the 2 coordinate system origins during the posture process.

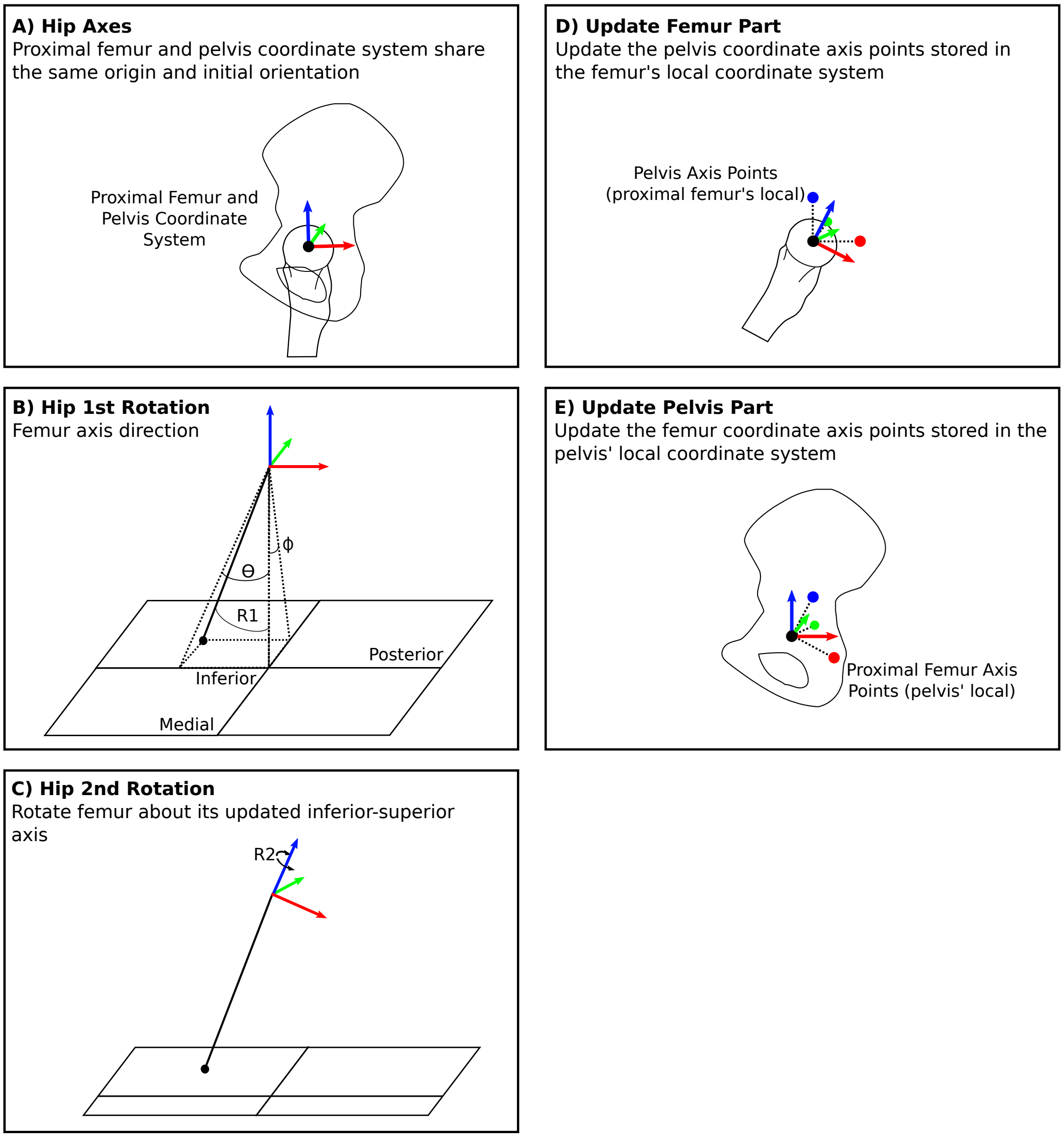

Fig. 5 The hip is defined as a ball and socket joint. The femur coordinate system rotates about the pelvis based on input anatomical angles.

The resulting posture is defined by angles for flexion-extension, abduction-adduction, and internal-external rotation. Flexion is defined as the angle between the reference and updated inferior-superior femur axis projected into the sagittal plane. Abduction-adduction is defined as the angle between the reference and updated inferior-superior femur axis projected into the coronal plane. Internal-external rotation is defined as the angle of rotation about the proximal femur's inferior-superior axis relative to its initial orientation.

The first rotation applies the flexion and abduction-adduction angles as shown in Fig. 5b. The pelvis coordinate system is held fixed during this part of the process. A point and vector that form the desired angles are calculated first. Then a rotation is applied to orient the inferior-superior axis of the femur along this direction. The second rotation in the process applies external-internal rotation as shown in Fig. 5c. The initial orientation has the anterior-posterior axis of the femur parallel with the sagittal plane. The femur coordinate system is then rotated about its updated inferior-superior axis by the prescribed number of degrees.

The last step of the function is to store the neighboring coordinate system locations. This is shown in Figs. 5d and 5e. The femur part stores pelvis coordinate system location in the femur local coordinate system. The pelvis part stores the location of the femur's coordinate system location in the pelvis local coordinate system.

3.4 Spine

The spine is composed of many vertebrae separated by an intervertebral disc that allows for relative motion between 2 neighboring vertebrae. Each vertebra pair could potentially be implemented into the current framework with its own joint function and inputs. However, this would require a large amount of user input to define a target posture and would not be an intuitive process. Therefore, the spine is treated as a special case, and a specialized function has been developed to define the spine position in any posture with minimal input requirements.

The spine process is illustrated in Fig. 6. The current spine model includes the thoracic and lumber vertebrae. The position of the vertebrae in the reference position has been approximated by a Bézier curve that passes through the center of each vertebra as shown in Fig. 6a. The curve has end points in the T1 vertebra and the sacrum with 2 control points to define the curvature. The tangent vector of the curve

at each of these points is approximately parallel to the inferior-superior axis of the vertebra. The spacial distribution of the vertebra relative to the total length of the curve is also calculated.

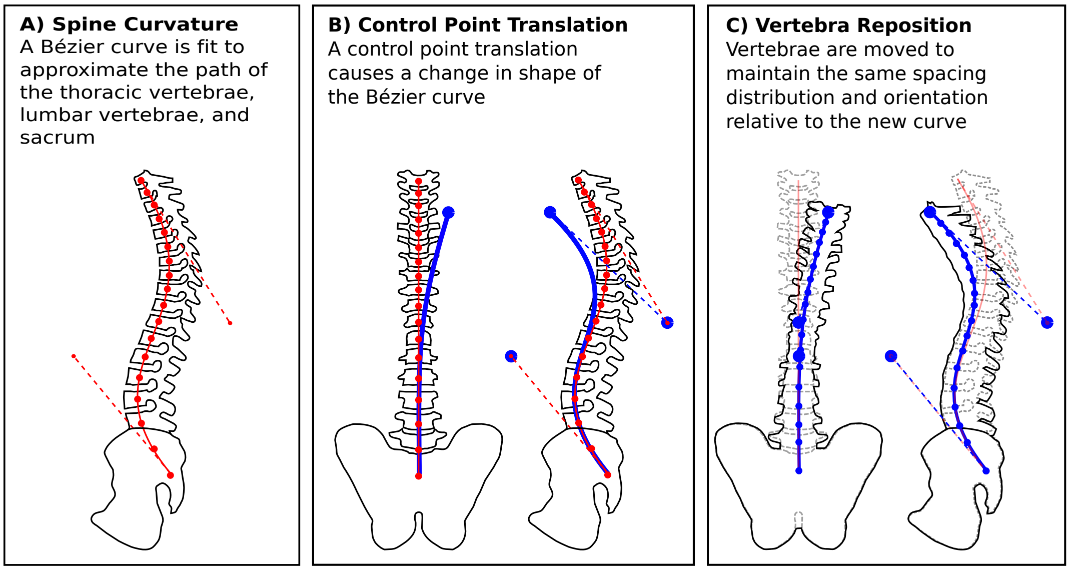

Fig. 6 The spine posture is defined by a Bézier curve. A translation is applied to the control points to adjust the posture of the spine without the need to manipulate each individual vertebra.

A new posture is defined by inputing displacement values for one or all of the points that define the Bézier curve. Figure 6b shows an updated curve in blue after the T1 end point has been translated in the inferior and medial directions. The vertebrae are repositioned on the new curve using the spacial distribution that was measured in the reference position. A rotation is then applied to each vertebra to align its inferior superior axis with the tangent vector of the updated curve. This rotation is currently performed as the minimal rotation to reach the new orientation.

4. Posture Assembly

The updated nodal coordinates of the bone geometries are built as an assembly of each individual bone rotated and translated by the joint functions. The posture of the assembly is specified by a total of 6 angles that are put into the joint functions. Two angles from the ankle joint, one angle from the knee, and 3 more at the hip. These 6 angles can be thought of as 6 new parameters in a state space that can now be smoothly varied to understand the effects of posture on injury. Alternatively, these 6 angles could be thought of as indices pointing to a catalog of FE models that

meet the posture requirements of an experimental setup. Thus a single FE model could be accompanied by a library of postures that it could be obtained in for quick calculations.

The process of creating the assembly is illustrated in Fig. 7 with Figs. 7a–c showing examples of posturing for the ankle, knee, and hip that would be performed in the joint functions. At these step, these postures are performed independently of each other. The assembly process begins with a single anchor point that is placed in the global coordinate system. This example uses the foot as the anchor point as shown in Fig. 7d, but the procedure allows for any bone to be used as the anchor. The foot contains the location of the tibia's coordinate system that was saved during the ankle function. This is converted from the foot's local coordinate system into the global coordinate system based on the foot's current position. This allows the tibia to be placed into the global coordinate system at this location as shown in Fig. 7e. The process continues in Fig. 7e where the tibia contains information on the location of the coordinate system of the patella and femur. These locations are converted from the tibia's local coordinate to the global coordinate system relative to the tibia's current position to place the femur in Fig. 7f. Finally, the pelvis coordinate system that is stored in the femur is output to place the pelvis into the global coordinate system in Fig. 7g.

The posture procedure also has the capability to combine the posture assemblies after posture has been applied to each relationship set separately. For example, the leg and spine can be run through the procedure independently and then combined into the same global coordinate system. This ability is necessary to accommodate for the spine's irregular posture process when attaching the spine to the legs. The assembly combination is performed by using a common part that is defined in each assembly. For the leg and spine, the sacrum can serve as the common part. The assemblies are combined by transforming the sacrum in one assembly to be in the same position and orientation as the sacrum of the other assembly. This is performed at each step in the posture procedure after the posture for both of the assemblies has been set.

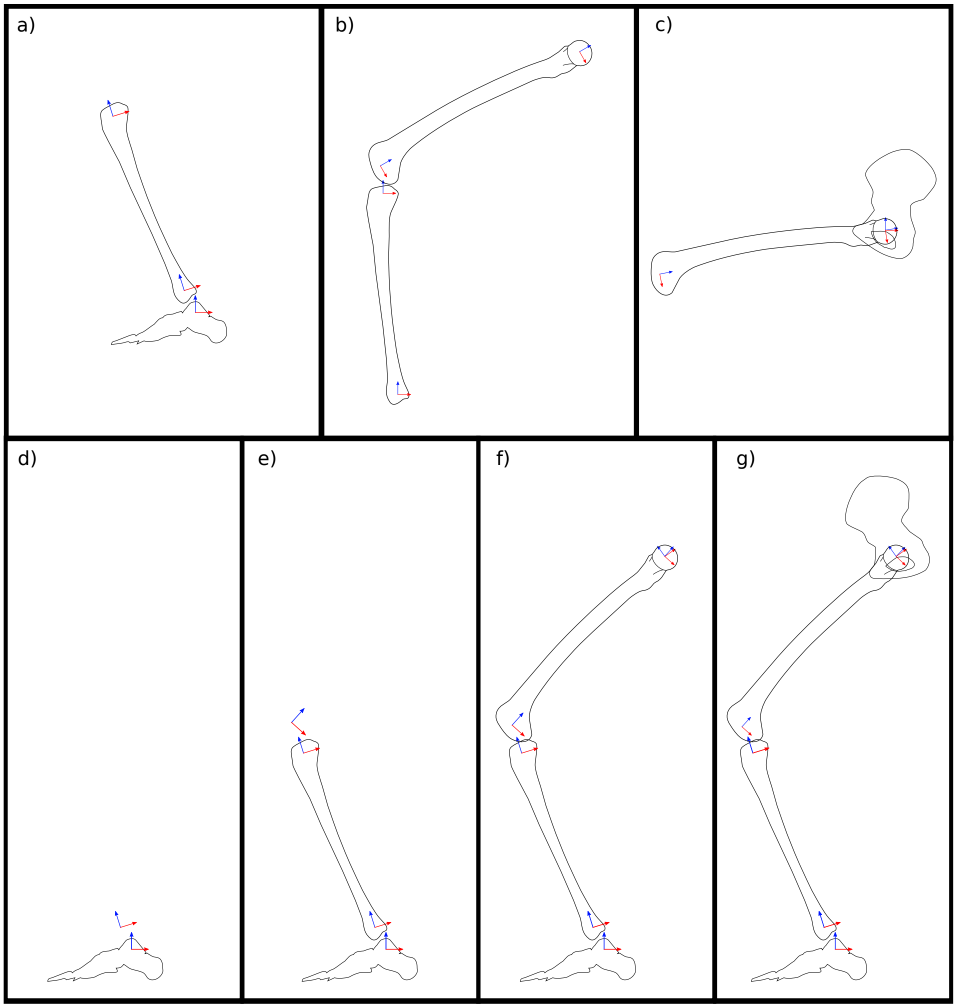

Fig. 7 The postured model is assembled by building from an anchor point using the stored coordinate system in each part. a–c represent independent posture functions. d–g show the assembly of the model in sequential steps.

5. Developing Load Curves for Prescribed Postures

Specifying the rigid-body motion of the bones within the assembly is done through a series of load curves. In general these are coordinated motions that include translations of the nodes that compose the bones in the assembly. Some care must be taken to ensure the load curve does not over determine the motion of the joint or impose a finite deformation of the presumed rigid bodies. A simple analogy comes from that of pendulum motion. Only considering final coordinates of the pendulum will lose the circular arc, thus a linear interpolation between the starting and ending point will result in compression and shear strains imposed on the pendulum in between. To ensure the motion of the bones remains rigid through-

out the simulation phase, the previously outlined procedure is repeated at small angle increments so that there are several steps from the initial posture to a final posture. Node data at each step are stored to obtain a set of geometries that will provide a smooth motion. The posture FE simulation is carried out in LS-DYNA. Three curves are defined for each node with the ∗DEFINE_CURVE keyword to contain the x, y, and z displacement. The displacement is applied using the ∗BOUNDARY_PRESCRIBED_MOTION_NODE keyword with the VAD option set to a value of 2 to indicate a displacement.

Load curves can be created to apply motion to joints sequentially, but they are currently created so that the final angle for all joints is reached at the end of the posturing simulation to minimize the total simulation time. Alternatives that still need to be explored are countless. It will be important to consider the inertia and deformation of the soft tissues that surround the bones. The model contains muscles and ligaments that connect bones across joints. Interactions between muscles, skin, and bone may result in a transformation that is not unique based on which bone is used to start the assembly process or the order that the joints are moved. It will be necessary to consider moving the joints sequentially rather than all at once to find a pattern that produces the best results. Inertia of the soft tissue is another concern that will need to be considered. The example used in Fig. 7 takes the foot as the starting bone and anchor point that is held in place throughout the posturing process. As a result, the muscles at the hips will have the highest velocity and inertia. It will be necessary to determine which areas of the leg are more susceptible to error in tissue deformation resulting from inertia. It may be necessary to make use of an anchor point that is not contained within any of the bones. An anchor point at the final posture's center of gravity could help to minimize inertia in the leg.

6. FE Simulations

The posture procedure defines the node locations of the bones in the leg that are required to create the desired posture. An explicit FE simulation is performed in LS-DYNA to apply a displacement to the nodes. The bone's translation applies force to the surrounding muscles, ligaments, and skin to move the leg to the desired posture. Material properties can be optimized to allow for faster simulation time because only the final node coordinates are important. Stress that is developed in the tissue during the posturing simulation is not used in future simulations. A list of material properties is shown in Table 1. These material properties provide a larger time step to allow for longer time periods to be simulated. Simulating a longer time period requires lower velocities to reach the target posture, which will reduce inertia and unwanted soft tissue deformation.

Table 1 Material properties are chosen to optimize simulation performance to reduce simulation time and prevent excessive deformation from inertia

Material	Density (kg/m^3)	Young's modulus (GPa)	Poisson's ratio
Muscle	1800	1.5	0.30
Ligament	1800	1.5	0.30
Skin	1000	1.0	0.49

An FE was performed where the leg moves from its initial position to 15° dorsiflexion, 85° knee flexion, and 85° hip flexion. A plot of the joint angles that makes up this motion is shown in Fig. 8. Here the dashed traces for knee rotations are dependent on the flexion angle. The relative knee translation is also dependent on the flexion angle and represents the translation of the femur coordinate system at the TEA relative to the tibia coordinate system. Figure 9 shows the change in geometry that takes place during the simulation.

Nodal coordinates for all parts of the model are output at steps throughout the simulation. These updated positions can then be used as the input mesh for subsequent simulations to make use of the new posture. This method allows a range of mesh postures to be output from a single simulation when moving from the initial to final posture.

During the posturing FE simulation, the patella protrudes through the flesh layer.

This represents an error introduced from a contact failure within the FE simulation. Improving the solver step or the underlying model will improve this result. We emphasize that this is separate and independent from the underlying joint function.

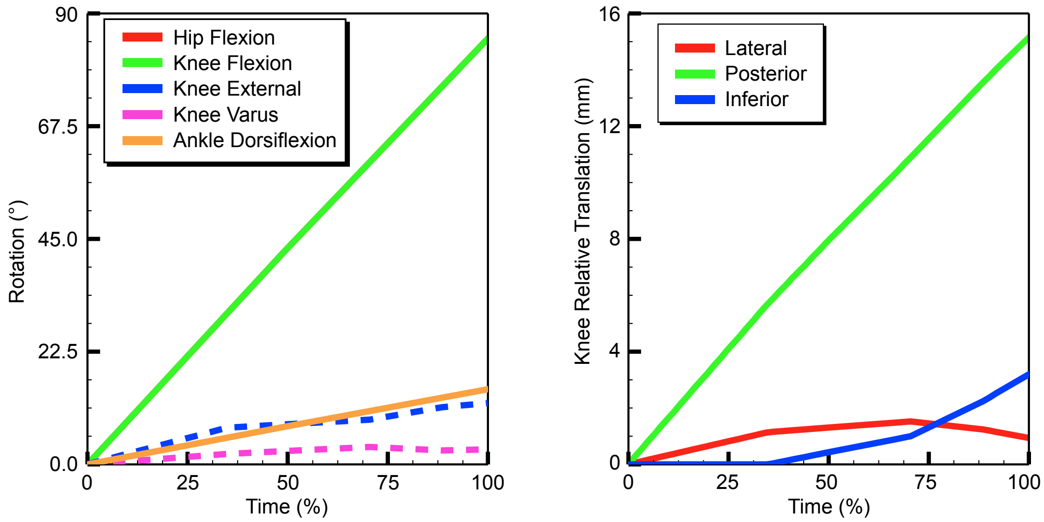

Fig. 8 A plot of the joint angles and knee translations that are input into the FE simulation. Dashed traces for rotation and the knee translation are coupled to knee flexion. Knee translation is the translation of the femur coordinate system at the TEA relative to the tibia.

There are several considerations and options that need to be explored for the FE simulation. The current method makes use of an explicit analysis. An implicit approach would remove concerns about the time step and inertia. However, the explicit method was chosen due to the large number of contact definitions in the leg model. There are also methods of connecting soft tissue to bone that will need to be investigated. This includes the possibility of using truss elements to tether the soft tissue and bones together. This could prevent excessive separation of tissues while still providing the ability to have relative motion that would be lost with a tied contact method.

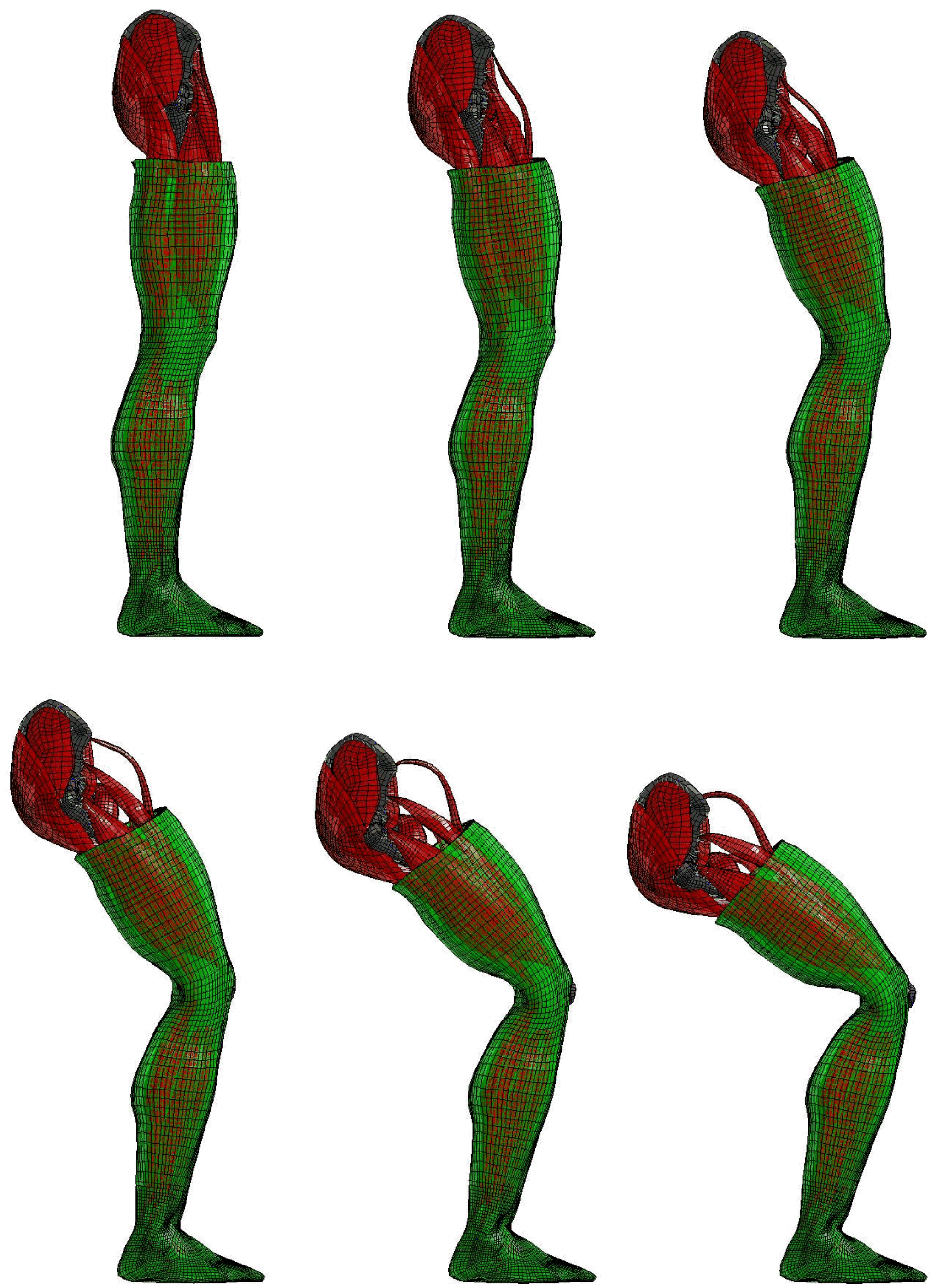

Fig. 9 An image montage of the leg being postured from a standing posture to a 15° ankle dorsiflexion, 85° knee flexion, and 85° hip flexion

7. Current Limitations

In addition to the considerations listed in the body of this text, it will also be important to acknowledge physiological changes in muscle tissue that cannot be captured through an FE simulation. A real human body moves due to contraction and relaxation of muscles attached to the skeletal system. This activity results in a change in cross sectional area and tension within the muscles. Muscles are passive in the current model, which could result in errors in the final geometry. A potential complication of this can be seen in Fig. 9. The sartorius muscle bends upwards near its origin at the pelvis as the hip flexion angle increases. This is the result of the distance between muscle connection points becoming shorter as the pelvis rotates forward rather than an effect of inertia. This problem can possibly be resolved by applying an artificial strain to muscles in their longitudinal direction to recreate the contraction that would take place in a real muscle.

However, other methods of adjusting the posture of a model will suffer from error in this aspect as well. Sculpting by hand is highly subjective for the geometry of an individual posture. It will also be difficult to maintain consistent procedures and interpretations of bone and relative movement of soft tissue over multiple postures. Re-scanning the body over multiple postures requires an intense effort to acquire many images and to process these images to create geometry and a FE mesh. In addition to the time investment, the large biological variability in both geometry and joint kinematics means that a model for a single set of imaging data can not represent a whole population. The posture procedure that is described in this report has the advantage of allowing for variability. Joint functions can be developed to describe a variable population more easily than performing an imaging study. Combined with efforts to capture geometric variability,[1] this procedure allows for consistent development of models that better represent a population.

8. Future Work

The procedure described here contains a generalized framework that was created with the intent to allow for updates and expanded capabilities. The current procedure is capable of positioning the leg, lumbar, and thoracic spine based on user input. However, more work is required to improve the accuracy and to extend this capability to other areas of the body. Future work will include improvements to accuracy of the joint functions, definitions for additional joints and rigid bodies, and

improvements to the FE procedures to apply the motion.

The ankle and hip are currently defined as simplified ball and socket joints. These simplified functions allow for a proof-of-concept display of the posture procedure but will require additional detail to fully capture the kinematics of these joints. These functions will be updated to make use of kinematic data similar to the knee function described in this work. The ankle in particular will be important to understand how the initial relative position of the calcaneus, talus, tibia, and fibula may affect the transfer of load from the foot into the long bones of the leg. Efforts to validate the posture process have been started by defining a boundary condition similar to a pendulum impact test in literature that compared the force response from a foot in a neutral position and dorsiflexion.[3] This effort may require a more detailed ankle function to be implemented to properly capture the transmission of force from the foot to the tibia. This effort will be detailed in future reports along with any updates that are required for the ankle function.

Improvements will also be made to the FE simulation procedures that apply the motion. Work will be done to examine which order of joint motion will produce the most realistic final results. The current method requires one of the bones in the model to remain fixed to be used as a reference point for motion in other bones. The capability to choose an arbitrary position as an anchor point of the simulation will be added. This option can be used to minimize inertia during the simulation. For example, the center of mass of the leg could be used as an anchor point to reduce the maximum velocity required to create the input motion. The use of mass proportional damping and the use of beam elements connecting soft tissue will also be explored to improve soft tissue deformation and relative movement during the motion. Additionally, other methods involving approximations of finite deformations are being pursued that might offer the removal of the dependence on an external FE simulation completely.

The FE simulation portion of the posture procedure requires the most time to run. A potential method to reduce the need for a FE simulation is the use of computationally empirically derived joint functions. After running an initial simulation for a joint, the motion of the soft tissue can be saved. These data would allow for that joint motion to be used and combined with other motions without the need to simulate it again. As an example, consider the process that would be necessary to have

a set of models that compare various combinations of knee and ankle rotations. Using simulations, it would be necessary to simulate the bending of one joint while the other is set to discrete angles. One simulation would be required for each discrete angle. However, with the use of a computationally empirically derived joint function, it would be possible to perform one simulation of knee flexion and one simulation of ankle flexion to then combine the results to create a continuous range of possible postures. This would also improve inertial effects if joints can be simulated independently and then be combined with other body regions to form a larger model.

9. Conclusion

The work presented in this report has demonstrated an initial capability to move a leg and spine model into a prescribed posture by defining the relative motion of bones at joints. The motion is applied to the model through a FE simulation. While more work is necessary to refine the process, this provides a framework with which to expand on.

This procedure adds to the suite of tools described previously[1] that can be used to capture biological variability in the leg model. These tools enable creation of models that can accurately represent the Soldier population for a range of different tasks and body positions. This ability is necessary to recreate experimental conditions to improve model validation efforts. The ability to posture is also necessary to create specialized models to assess a large range of threats without the need to mesh a geometry for each situation. Eliminating the meshing step allows research efforts to focus on simulations that can provide a greater understanding of injury risk and assist with the design of protection solutions.

10. References

1. Sokolow A. Progress in finite element modeling of the lower extremities. Aberdeen Proving Ground (MD): Army Research Laboratory (US); 2015 Jun. Report No.: ARL-MR-0890.

2. Li G, Papnnagari R, Nha K, DeFrate L, Gill T, Rubash H. The coupled motion of the femur and patella during in vivo weightbearing knee flexion. Journal of Biomechanical Engineering. 2007;129(6):937–943.

3. Gallenberger K, Yoganandan N, Pintar F. Biomechanics of foot/ankle trauma with variable energy impacts. Annals of Advances in Automotive Medicine. 2013;57:123.

INTENTIONALLY LEFT BLANK.

Appendix. Knee Kinematic Data

The knee function makes use of kinematic data in a lookup table that has been reported in literature.[1] All kinematic parameters for the femur and patella in relation to the tibia are dependent of the knee's flexion angle. However, there is variability on joint kinematics that results from geometric variability. As a result, the knee used in this model deviates from the average values reported in literature for some parameters. The lookup table used for the femur-tibia relationship is shown in Table A-1. The patella-tibia relationship is shown in Table A-2.

[1]Li G, Papnnagari R, Nha K, DeFrate L, Gill T, Rubash H. The coupled motion of the femur and patella during in vivo weightbearing knee flexion. Journal of Biomechanical Engineering. 2007;129(6):937–943.

Table A-1 Femur lookup table values used in the posture procedure. The directions listed are considered the positive directions.

Rotation (°)			Translation (mm)		
Flexion	Internal rotation	Varus	Posterior	Medial	Superior
0.0	0.0	0.0	0.0	0.0	0.0
4.7	–1.1	0.3	0.9	–0.2	0.0
9.3	–2.3	0.6	1.8	–0.4	0.0
14.0	–3.4	1.0	2.7	–0.5	0.0
18.6	–4.5	1.3	3.5	–0.7	0.0
23.3	–5.7	1.6	4.4	–0.9	0.0
27.9	–6.8	1.9	5.3	–1.1	0.0
32.6	–7.5	2.2	6.2	–1.2	0.1
37.2	–7.7	2.4	7.0	–1.2	0.4
41.9	–8.0	2.6	7.8	–1.3	0.7
46.6	–8.2	2.9	8.6	–1.4	1.0
51.2	–8.5	3.1	9.4	–1.4	1.2
55.9	–8.7	3.3	10.2	–1.5	1.5
60.5	–9.0	3.5	11.0	–1.5	1.8
65.2	–9.8	3.2	11.8	–1.4	2.5
69.8	–10.6	3.0	12.6	–1.3	3.1
74.5	–11.3	2.8	13.5	–1.2	3.8
79.1	–11.8	2.9	14.3	–1.1	4.6
83.8	–12.2	3.0	15.0	–1.0	5.4
88.4	–12.5	3.1	15.8	–0.8	6.1
93.1	–12.6	3.8	16.4	–0.7	7.1
97.8	–12.5	4.7	16.9	–0.7	8.2
102.4	–12.4	5.7	17.4	–0.6	9.2
107.1	–12.3	6.9	17.9	–0.5	10.3
111.7	–12.3	8.3	18.4	–0.4	11.3
116.4	–12.3	9.7	18.9	–0.4	12.3
121.0	–12.2	11.0	19.4	–0.3	13.4
125.7	–11.6	12.1	19.9	–0.3	14.6
130.3	–11.1	13.2	20.4	–0.4	15.7
135.0	–10.5	14.3	20.9	–0.4	16.9

Table A-2 Patella lookup table values used in the posture procedure. The directions listed are considered the positive directions.

	Rotation (°)			Translation (mm)		
Knee flexion	Patella flexion	Medial tilt	Lateral rotation	Posterior	Medial	Superior
0.0	0.0	0.0	0.0	0.0	0.0	0.0
4.7	1.7	–1.3	0.2	0.7	–0.9	1.0
9.3	3.5	–2.6	0.4	1.3	–1.8	2.1
14.0	5.2	–3.9	0.6	2.0	–2.8	3.1
18.6	7.0	–5.2	0.7	2.7	–3.7	4.1
23.3	8.7	–6.5	0.9	3.4	–4.6	5.2
27.9	10.5	–7.8	1.1	4.0	–5.5	6.2
32.6	12.0	–8.7	1.4	4.7	–6.2	6.7
37.2	13.2	–9.3	1.7	5.4	–6.6	6.9
41.9	14.5	–9.9	2.0	6.0	–7.0	7.0
46.6	15.8	–10.6	2.3	6.7	–7.4	7.1
51.2	17.1	–11.2	2.5	7.4	–7.8	7.2
55.9	18.4	–11.8	2.8	8.0	–8.3	7.4
60.5	19.6	–12.5	3.2	8.7	–8.7	7.5
65.2	20.4	–13.6	3.6	9.7	–9.3	7.3
69.8	21.3	–14.7	4.1	10.6	–9.9	7.2
74.5	22.2	–15.8	4.6	11.5	–10.5	7.1
79.1	22.9	–16.0	4.5	12.3	–10.5	6.9
83.8	23.7	–16.0	4.3	13.2	–10.5	6.7
88.4	24.5	–16.0	4.0	14.0	–10.4	6.5
93.1	25.4	–16.0	3.8	14.7	–10.2	6.0
97.8	26.4	–16.1	3.6	15.4	–9.9	5.3
102.4	27.4	–16.1	3.4	16.0	–9.6	4.6
107.1	28.6	–16.0	3.2	16.6	–9.3	3.9
111.7	30.0	–15.6	2.9	17.3	–8.9	3.2
116.4	31.4	–15.1	2.6	17.9	–8.5	2.4
121.0	32.8	–14.5	2.3	18.5	–8.1	1.7
125.7	34.2	–13.2	1.7	18.9	–7.6	1.0
130.3	35.6	–12.0	1.1	19.3	–7.2	0.3
135.0	37.0	–10.7	0.5	19.8	–6.7	–0.4

1 (PDF) DEFENSE TECHNICAL
INFORMATION CTR
DTIC OCA

2 (PDF) DIRECTOR
US ARMY RESEARCH LAB
RDRL CIO L
IMAL HRA MAIL & RECORDS MGMT

1 (PDF) GOVT PRINTG OFC
A MALHOTRA

59 (PDF) DIR USARL
RDRL DPW
R COATES
P FROUNFELKER
RDRL ROP L
F GREGORY
RDRL SL
N EBERIUS
RDRL SLB G
N ELDREDGE
RDRL SLB W
D BOOTHE
P GILLICH
C KENNEDY
A KULAGA
W MERMAGEN
RDRL WM
S KARNA
S SCHOENFELD
RDRL WML C
T PIEHLER
RDRL WML H
B SCHUSTER
RDRL WMM B
G GAZONAS
B LOVE
R WILDMAN
RDRL WMM D
M LYNCH
RDRL WMM G
L PIEHLER
N ZANDER
RDRL WMP A
S BILYK
RDRL WMP B
S ALEXANDER
A DAGRO
A DILEONARDI
A EIDSMORE
W EVANS
C GUNNARSSON
C HAMPTON
C HOPPEL
Y HUANG
M KLEINBERGER
J MCDONALD
P MCKEE
S SATAPATHY
A SOKOLOW
K THOMPSON
C WEAVER
T WEERASOORIYA
S WOZNIAK
T ZHANG
RDRL WMP C
R BECKER
T BJERKE
D CASEM
J CLAYTON
M GREENFIELD
B LEAVY
J LLOYD
A TONGE
RDRL WMP D
R DONEY
J RUNYEON
RDRL WMP E
P SWOBODA
RDRL WMP F
E FIORAVANTE
N GNIAZDOWSKI
R GUPTA
R KARGUS
M TEGTMEYER
RDRL WMP G
R BANTON
S KUKUCK
R SPINK

INTENTIONALLY LEFT BLANK.